Biomechanical Evaluation of Military Aviation-Related Occupational Tasks on the Spine

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1. Introduction

Work-related low back disorders and neck pain are becoming prevalent medical issues among both the civilian and military populations. On a worldwide scale, over 540 million people experience low back pain (1). In the United States Military, low back disorders are one of the most common reasons for seeking medical attention (2). Of the soldiers in Operation Iraqi Freedom who suffered low back injuries, 86% never returned to active duty (2,3,4). Studies have shown that servicewomen have increased rates of low back pain than their male counterparts (3). As the percentage of women in the military increases, this is developing into an acute issue. Compared to the general population, military members experience neck pain at a greater rate due to their occupational environment (5). Combat helmets and reliance on heads-down devices, lead to increases in neck kinematics which contribute to pain (6). Due to the high occupational demands of soldiers, any amount of back or neck pain may compromise a mission and can sideline highly trained asset for an extended or indefinite period.

The field of exoskeletons is rapidly developing to better address and prevent low back disorders and neck pain. These wearable devices are designed to relieve the load on the wearer by providing support or transferring the applied forces. Exoskeletons come in both passive and active forms. Passive exoskeletons operate through mechanical devices such as springs, cables, or other materials. Active exoskeletons usually contain motors which create force to allow the user to lift objects much heavier than normally possible. Although these devices have the potential to reduce the number of low back injuries in the workplace, sparse research has been conducted to investigate their effectiveness in military environments.

The objective of this study was to quantify potential tradeoffs of two passive upper extremity exoskeletons on the lumbar spine.

2. Methods

2.1 Approach

In this laboratory study, two commercially available exoskeletons were evaluated and compared to a non-exoskeleton control condition. The exoskeletons in this study are classified as upper extremity exoskeletons. This model of exoskeleton is designed to be worn in environments where the user’s arms are primarily extended in front of their chest or above their heads.

Image 1: Levitate Airframe

Image 2: EksoBionics Eksovest

An EMG assisted dynamic biomechanical spine model, which has been proven and employed (7), was implemented to understand the biomechanical impact of the exoskeletons. An overview of the data collected includes the muscle forces of key power-producing trunk muscles(via EMG), the lateral shear, A/P shear, and compression of the lumbar intervertebral discs (spine model), full body kinematics (via motion capture), and ground reaction forces (via force plate). Additional inputs of the dynamic biomechanical spine model include MRI-derived muscle locations and sizes, subject-specific anthropometry, and tissue material properties.

2.2 Subjects

8 male subjects were recruited to participate in this study. The subjects had a mean (SD) age of 20.7 (1.5) years, stature 182.1 (2.7) cm, and mass 88.2 (14.8) kg. All subjects provided informed consent and self-reported no history of low back conditions or pain requiring medical attention.

2.3 Study Design

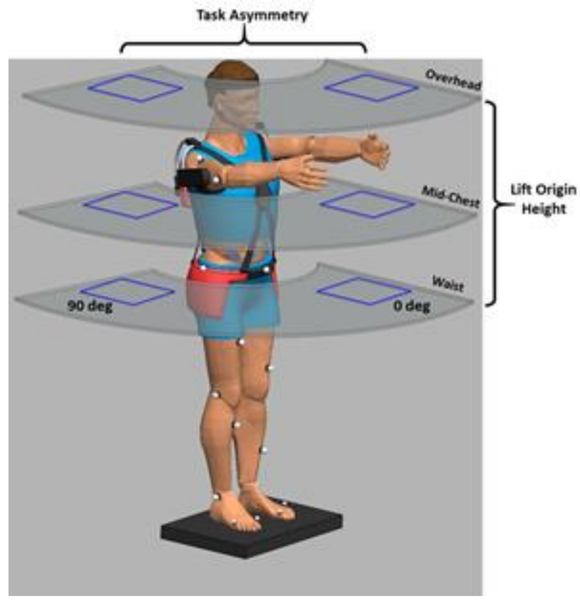


Image 3: Task Height & Asymmetry

A balanced 3 x 3 x 2 x 2 x 2 mixed model design was applied to this study. Independent variables include intervention (wearing of exoskeleton), lift origin/task height (waist, mid-chest, overhead), task asymmetry, and repetition. Two separate tasks (lifting and torqueing) were completed by each subject to simulate the condition in a military workplace environment. The lifting task consisted of lifting a 11.34 medicine ball while the torqueing task included screwing three screws with a torque wrench of set output of 78 N*m. The order of the trials performed by each subject was first blocked by intervention then task type then asymmetry. These three blocks were counterbalanced for each subject to reduce the possibility of order or fatigue effects.

2.4 Instrumentation

EMG data was bilaterally collected for the 10 power producing muscles of the trunk. These include the latissimus dorsi, erector spinae, internal oblique, and rectus abdominis. EMG data was collected at 1000 Hz using bipolar surface electrodes and a wireless Trigno system (Delsys, Natick, MA, USA). The signals were then notch filtered at 60 Hz and band-pass filtered at 30-450 Hz. A fourth order low pass filter was then used to rectify and smooth the signal with a cutoff frequency of 1.59 Hz and time constant of 100 ms. The subject kinematics were recorded with a 42-camera Prime 41 OptiTrack optical motion capture system (NaturalPoint, Corvallis, OR, USA) at a 120 HZ sampling rate. During the trials, subjects stood on a 6090-15 six-axis force plate (Bertec, Worthington, OH, USA) recording the ground reaction forces during the length of each trial. All signals were simultaneously collected using custom laboratory software developed in Matlab (MathWorks, Natick, MA, USA) and synced with a data acquisition board (USB-6225, National Instruments, Austin, TX, USA).

2.5 Testing Procedure

Upon arrival to the laboratory, each subject was informed of the goals of the study and the data collection procedure. Once providing their written consent, the subjects anthropometric data was recorded. This included their height, weight, waist circumference, and the breadth and width of the torso at the sternum and umbilicus levels. The subjects were then outfitted with EMG sensors placed along 5 core trunk muscles, erector spinae, external obliques, internal obliques, latissimus dorsi, and rectus abdominis. An additional EMG sensor was placed across the chest to measure cardiac activity. Forty-one reflective optical motion capture markers were next placed on the subject in accordance with the guidelines of OptiTrack's motion capture software. Additional markers were placed on the force plate. Once sensor placement was finalized, the model was calibrated for each subject using data derived from a specific set of dynamic concentric and eccentric lumbar motions while holding a 9.07kg medicine ball (7).

Next the lifting and torqueing trials were performed by the subject. The intervention and order of task followed the study design previously mentioned. Prior to starting, the subject's feet locations were marked on the force plate to ensure consistent placement among all trials. The lift origin/task height was marked for each height on each of the testing structures for lifting and torqueing. The distance of the structure away from the subject was also marked on the floor to ensure consistency. All tasks for one exoskeleton were performed consecutively. With three interventions, each exoskeleton was fitted to the subject directly before performing the tasks for that particular exoskeleton. As mentioned earlier, the order of tasks was blocked off by task type (lifting or torqueing) and asymmetry. The lift origin/task height was randomized among each subject.

2.6 Data Analysis

The dependent variables were calculated via simulations in Adams (MSC Software, Santa Ana, CA, USA), a multibody dynamics solver, using the EMG-driven lumbar spinal model. These results were then extracted an analyzed with JMP Pro 13 software (SAS Institute Inc., Cary, NC, USA). To asses the effects of independent measures and two-way interactions a 2-way analysis of variance (ANOVA) was performed. Post-hoc analyses were then implemented using a Tukey HSD test where appropriate.

3. Results

3.1 Lifting Task

The output of the biodynamic spine model estimated the peak average magnitude of spinal loading for compression, A/P shear, and lateral shear. The peak compression value was found to be at the L4/L5 spinal level, while the A/P shear and lateral shear values were found at the L2/L3 and L5/S1 spinal levels, respectively. Any differences found between the control condition and either intervention were negligible compared to the overall magnitude of loading for both compression and A/P shear. The magnitude of lateral shear values was generally low across all experimental conditions compared to the values of compression and A/P shear and thus showed marginal biomechanical significance.

Using the aforementioned 2-way ANOVA, the significance of interaction effects was able to be determined. There was found to be a statistically significant main effect of exoskeleton observed for peak lateral shear ($p<0.01$). Another significant main effect of height was observed for peak A/P shear ($p<0.001$). Additionally, for peak A/P shear, there was a significant asymmetry * height interaction effect observed ($p<0.01$). The final interaction effect observed for the lifting task was a significant exoskeleton * asymmetry effect for peak lateral shear ($p<0.001$).

Lifting Tasks	**Exoskeleton (E)**	**Task Height (H)**	**Asymmetry (A)**	**E*H**	**E*A**	**H*A**
Compression (L4/L5 Inferior)						
A/P Shear (L2/L3 Superior)		***				**
Lateral Shear (L5/S1 Superior)	**				***	
Torqueing Tasks	**Exoskeleton (E)**	**Task Height (H)**	**Asymmetry (A)**	**E*H**	**E*A**	**H*A**
Compression (L4/L5 Inferior)			*			
A/P Shear (L2/L3 Superior)						
Lateral Shear (L5/S1 Superior)			**		**	

* denotes $p<0.05$, ** denotes $p<0.01$, *** denotes $p<0.001$

Table 1: Significant Interaction Effects

The other independent variables assessed in this study impacted the peak spinal loads in a somewhat predictable manner. The peak compression only noticeable increased at the overhead lift origin (21.3%) compared to the waist and midchest heights. The A/P shear gradually increased as lift origin elevated, 14.6% at midchest and 41.0% at overhead. Lateral shear showed no significant change with lift origin. The asymmetric condition greatly increased lateral shear by 214.4%, while having no significant effect on compression and A/P shear. For peak compression and A/P shear, neither exoskeleton showed a significant difference compared to the control condition for any lift origin or asymmetry. Both exoskeletons decreased the lateral shear in the asymmetric position.

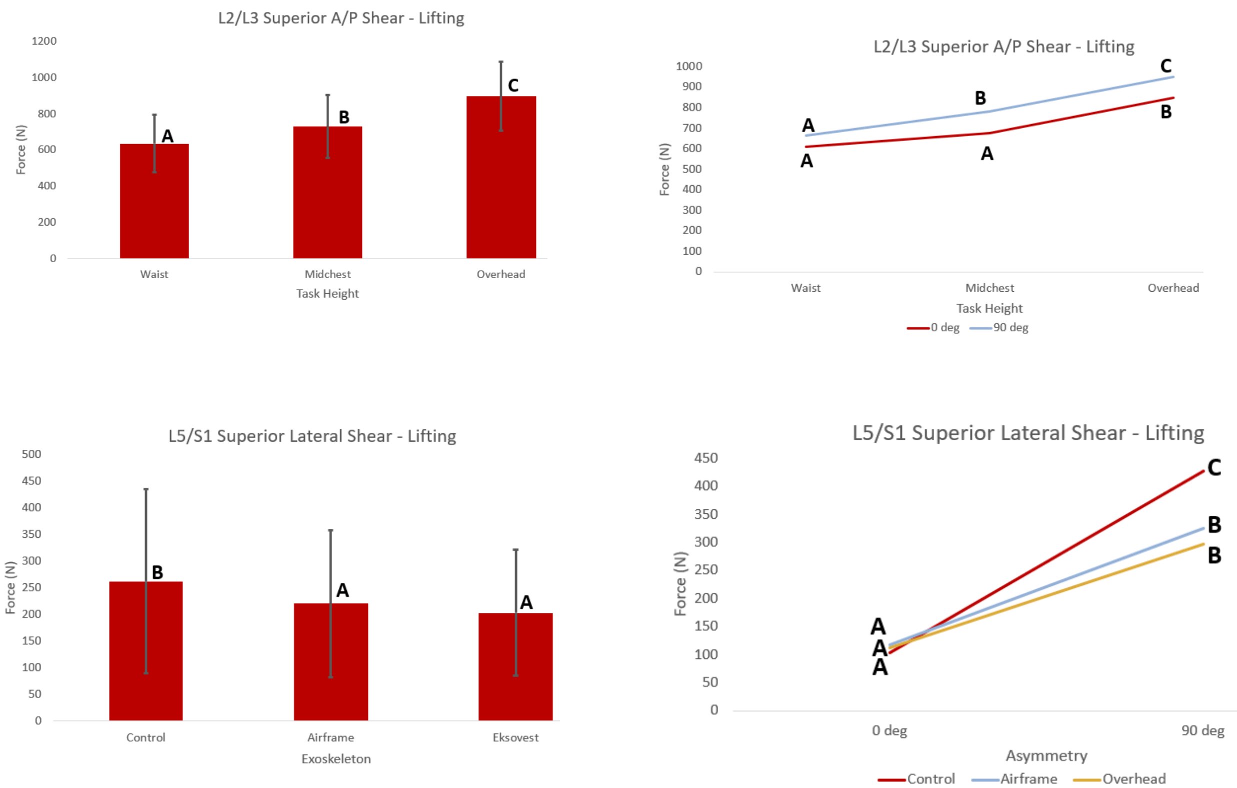
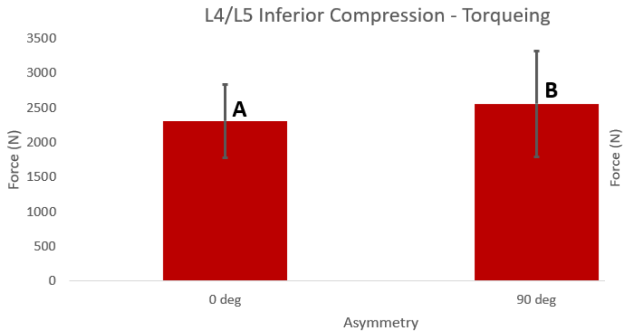


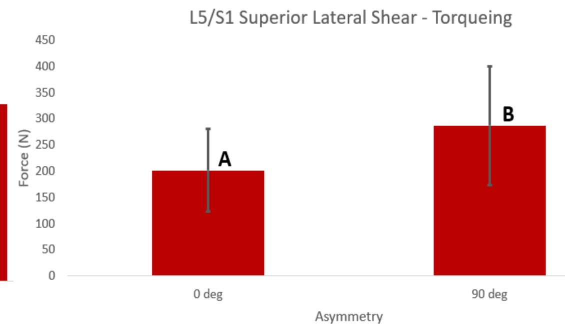
Figure 1: Peak lifting spinal loads with relevant interaction effects.

Levels not connected by the same letter are significantly different

3.2 Torqueing Task

The biomechanical results of the torqueing task were evaluated in a similar fashion to the lifting task. There was not found to be any statistical significance of exoskeleton for any loading direction. The effect of task asymmetry was determined to be significant for both compression ($p<0.05$) and lateral shear ($p<0.01$). Lastly, a significant exoskeleton * asymmetry interaction ($p<0.001$) effect was identified for the lateral shear.





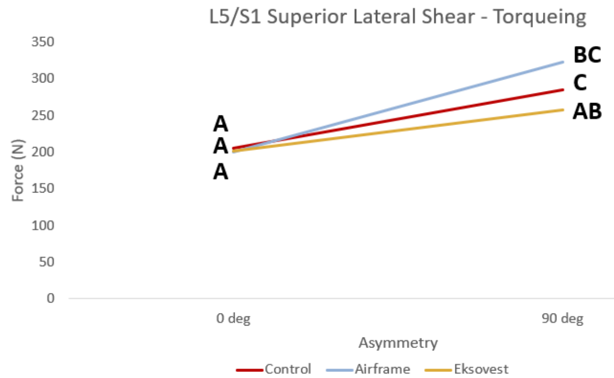


Figure 2: Peak torqueing spinal loads with relevant interaction effects. Levels not connected by the same letter are significantly different

The general trends of the data shown for the lifting task were not directly replicated for the torqueing task. Peak compression (-8.6%) and lateral shear (-27.6%) gradually decreased as the task height elevated while A/P shear increased (20%), percentage values shown display change between waist and overhead conditions. Loading in all directions increased with asymmetry with lateral shear increasing the most (29%).

4. Discussion

This study found that neither exoskeleton has any detrimental biomechanical effect on the user. The only directional loading that appears to be reduced by wearing an exoskeleton is the lateral shear, especially when used in the asymmetric position. The compression and A/P shear values were mainly affected by task height and asymmetry.

It is important to note that there is a discrepancy between the intention of the exoskeletons and the purpose of the biodynamic spine model. The model used in this study was designed to measure the spinal loads of the lumbar spine, while the exoskeletons were designed to redistribute shoulder loads directly to the hips. Therefore, the direct effects of the intervention were not studied, only the potential tradeoffs. The results of this study suggest that the exoskeletons were successful in redistributing the shoulder load, but this is not confirmed. On a positive note, the results indicate there is no negative repercussion on the loading of the lumbar spine.

As with any study, limitations were present in this investigation. Only 8 college-aged male subjects were tested, which is not representative of the military population. This small sample size limits the statistical power of the study and its ability to make definitive claims. The statistical analysis also may have been unable to detect significant differences that do indeed exist, especially interaction effects. A source of error may have come from the fit of the Eksovest. Unlike the Airframe which had multiple sizes and strength of should support spring, the laboratory only had access to one size of the Eksovest. This led to smaller and slimmer subjects not having the proper fit for maximum support. For these subjects, both the waist and arm bands were too large and not ideally secured.

Additionally, all trials took place in a controlled laboratory setting with untrained and inexperienced subjects. The subjects were not allowed to develop their own lifting or torqueing techniques, move their feet for the asymmetric tasks, and were clearly instructed on how to perform each task. Without instruction, the subject kinematics would have likely showed more variation. Future work of this study should include data collection on more subjects. With an increase in subjects, the statistical power of the results will also increase. Researchers should take actions to have the overall demographics of the subject pool reflect the gender and age of the military population. Additionally, the proper sizing of the exoskeletons should be ensured by obtaining smaller sizes of the Eksovest.

5. Conclusion

The results of this study imply that the use of either of the two upper extremity exoskeletons tested is not associated with a tradeoff by way of increased lumbar spine loading. Both exoskeletons performed similarly to the control condition in their handling of compression and A/P shear. The addition of an intervention did decrease lateral shear, but the value is generally low in magnitude such that it is only marginally significant. For the torqueing tasks an ANOVA analysis showed significant main effects of asymmetry for both compression and lateral shear, along with an interaction effect of exoskeleton*asymmetry for lateral shear. The lifting tasks the

A/P shear was found to be significant with respect to a height main effect and height*asymmetry interaction effect. The lifting lateral shear had an exoskeleton main effect and an exoskeleton*asymmetry interaction effect. This study only evaluated the potential spinal tradeoffs of these upper extremity exoskeletons and found there no be none. Changes in shoulder loading were not considered in this study but should be investigated in a future study. Due to the smaller unrepresentative sample size, this investigation possesses low statistical significance, a continuation of this study with more subjects will solidify the results.

References

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